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13. ABSTRACT (Maximum 200 words) A novel pulsed cooling paradigm (PCskin) integrating mean skin temperature (Tsk) feedback was compared with constant cooling (CC) or time-activated pulsed cooling (PC). Methods: Eight males exercised while wearing personal protective equipment (PPE) in a warm, dry environment (dry bulb temperature: 30 C; dew-point temperature: 11 C) in each of the tests. Treadmill exercise was performed (~225 w l m-2) for 80 min. A liquid cooling garment (LCG) covered 72% of the body surface area. Core temperature (Tc), local skin temperatures, heart rate, inlet and outlet LCG perfusate temperatures, flow, and electrical power to the LCG and metabolic rate were measured during exercise. At 75 min of exercise Tsk was higher (33.9 ± 0.5 C) or CC (32.0 ± 0.6 C) and PC > CC. The changes in Tc and heart rate during the tests were not different. Tc at 75 min was not different among the cooling paradigms (37.6 ± 0.3 C in PCskin, 37.6 ± 0.2 C in PC and 37.6 ± 0.2 C in CC). Heart rate averaged 124 ± 10 bpm in PCskin, 120 ± 9 bpm in PC and 117 ± 9 bpm in CC. Total body insulation (C . W1 . m-2) was significantly reduced in PCskin (0.020 ± 0.003) and PC (0.024 ± 0.004) from CC (0.029 ± 0.004). Electrical power in PCskin was reduced by 46% from CC and by 28% from PC. Real-time Tsk feedback to control cooling optimized LCG efficacy and reduced electrical pwoer for cooling without significantly changing cardiovascular strain in exercising men wearing PPE.				
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Skin Temperature Feedback Optimizes Microclimate Cooling

LOU A. STEPHENSON, CARRIE R. VERNIEUW,
WALIDA LEAMMUKDA, AND MARGARET A. KOLKA

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Introduction: A novel pulsed cooling paradigm (PC_{skin}) integrating mean skin temperature (T_{sk}) feedback was compared with constant cooling (CC) or time-activated pulsed cooling (PC). **Methods:** Eight males exercised while wearing personal protective equipment (PPE) in a warm, dry environment (dry bulb temperature: 30°C; dew-point temperature: 11°C) in each of the tests. Treadmill exercise was performed ($\sim 225 \text{ W} \cdot \text{m}^{-2}$) for 80 min. A liquid cooling garment (LCG) covered 72% of the body surface area. Core temperature (T_c), local skin temperatures, heart rate, inlet and outlet LCG perfusate temperatures, flow, and electrical power to the LCG and metabolic rate were measured during exercise. **Results:** At 75 min of exercise T_{sk} was higher ($33.9 \pm 0.2^\circ\text{C}$) in PC_{skin}, than in PC ($33.1 \pm 0.5^\circ\text{C}$) or CC ($32.0 \pm 0.6^\circ\text{C}$) and PC > CC. The changes in T_c and heart rate during the tests were not different. T_c at 75 min was not different among the cooling paradigms ($37.6 \pm 0.3^\circ\text{C}$ in PC_{skin}, $37.6 \pm 0.2^\circ\text{C}$ in PC and $37.6 \pm 0.2^\circ\text{C}$ in CC). Heart rate averaged 124 ± 10 bpm in PC_{skin}, 120 ± 9 bpm in PC and 117 ± 9 bpm in CC. Total body insulation ($^\circ\text{C} \cdot \text{W}^{-1} \cdot \text{m}^{-2}$) was significantly reduced in PC_{skin} (0.020 ± 0.003) and PC (0.024 ± 0.004) from CC (0.029 ± 0.004). Electrical power in PC_{skin} was reduced by 46% from CC and by 28% from PC. **Discussion/Conclusion:** Real-time T_{sk} feedback to control cooling optimized LCG efficacy and reduced electrical power for cooling without significantly changing cardiovascular strain in exercising men wearing PPE.

Keywords: body temperature regulation, liquid cooling garment, heat balance, personal protective equipment

AN EARLY APPROACH to extend the safe work duration for soldiers or workers wearing personal protective clothing (PPE) in a moderate or hot environment was to provide liquid cooling in a personal garment (18). Originally, the circulating perfusate was constantly cooled (CC) at a relatively low inlet temperature. However, the large power requirement for CC and the low perfusate temperature motivated us to investigate other approaches. We took a physiologic, integrative approach to effective personal cooling that would reduce the power requirement without significantly raising cardiovascular or thermoregulatory strain.

Skin temperature is an extremely important control for cutaneous vasomotor responses and the relationship between the two has been well-studied in humans. To summarize these findings: 1) skin temperature below thermoneutral ($\sim 33^\circ\text{C}$) activates vasoconstrictor signal(s) (8,10); 2) for skin temperatures at, and slightly above thermoneutral temperature ($\sim 33^\circ\text{C}$), the drive for vasoconstrictor activity is inhibited (1,10,11,13,17,19,24); and 3) for warm

or hot skin temperatures, there is frank vasodilation of cutaneous vasculature (10–12,23,24). These findings were instrumental in the development of our research paradigms. We reasoned that the key was to maintain skin temperature above that observed at thermoneutrality for rest so that the cutaneous vasculature remained minimally vasoconstricted, or even better, vasodilated at a level that supported rapid convective and radiative heat loss when cool liquid circulated in the cooling garment. This design would provide a microenvironment appropriate for heat transfer away from the body. We anticipated that further efficiency in cooling could be obtained by using pulsatile cooling that was controlled by temperature sensors placed on the skin so that real-time mean skin temperature could be used to activate the perfusate-circulating pump.

We hypothesized that: 1) pulsed cooling based on skin temperature feedback would be as effective in cooling the participants as determined by cardiovascular strain [which we define here as the change in heart rate and core temperature (T_c) over the exercise period] as constantly delivered cooling or pulsed cooling that was time-activated in the sequence of 2 min on and 2 min off; 2) pulsed cooling based on skin temperature feedback would result in a similar core temperature at the end of exposure as constantly delivered cooling or time-activated pulsed cooling; and 3) pulsed cooling based on skin temperature feedback would require less electrical power than constantly delivered cooling or time-activated pulsed cooling.

METHODS

Subjects: Eight men volunteered to participate in this IRB-approved study after all procedures and risks were explained both orally and in writing prior to obtaining their written informed consent. Due to the clothing and equipment worn in the experiments only military vol-

From the U.S. Army Research Institute of Environmental Medicine, Thermal and Mountain Medicine, Natick, MA.

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Address reprint requests to: Lou A. Stephenson, Ph.D., Thermal and Mountain Medicine, USARIEM, 15 Kansas St., Bldg. 42, Natick, MA 01760.

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unteers participated and female military volunteers were not available. Based on our earlier studies (14,15), these findings are relevant to both genders. Prior to testing, participants were medically cleared and then familiarized with all experimental techniques. Their average physical characteristics were: age, 20.6 ± 1.7 yrs.; height, 1.75 ± 0.05 m; mass, 73.14 ± 7.45 kg; DuBois surface area (4), 1.9 ± 0.11 m²; and BMI, 24.7 ± 1.3 ND.

Liquid cooling garment (LCG)/protective clothing: Subjects were fitted with a three-piece LCG that covered the head (hood), torso (vest) and legs (pants) prior to testing. The LCG design consisted of cotton or Nomex® (DuPont, Richmond, VA) aramid fabric woven or laminated around small diameter Tygon® (Saint-Gobain, Paris, Franc) tubing (2.5 mm, I.D.) divided into multiple parallel circuits. Total body surface covered was 72% (head = 6%, torso = 22%, and legs = 44%) as measured previously (3) using a Cyberware® (Cyberware, Inc., Monterey, CA) three-dimensional head and whole body scanner. The LCG was connected to a chiller circulator through umbilical tubes exiting the LCG at the waist (torso and legs) or collar (head).

Subjects wore a chemical protective clothing system over the LCG that included a charcoal-impregnated overgarment (top and bottom), cotton glove liners, butyl gloves, and M-40 chemical-biological field mask with hood and athletic shoes. Based on copper manikin studies, this configuration provided approximate insulative (C_{lo}) and vapor permeability (i_m) characteristics of 2.1 and 0.32, respectively (15).

Experimental design: Subjects completed three tests differing only in the cooling paradigm used. The three tests were: 1) constant cooling to all three regions (CC); 2) pulsed cooling to all regions with the circulating pump turned on or activated for 2 min, then deactivated for 2 min with this repeating cycle through out the experiment (PC); and pulsed cooling to all regions based on mean skin temperature feedback in which the circulating pump was activated when \bar{T}_{sk} reached 34.5°C, and deactivated when \bar{T}_{sk} reached 33.5°C (PC_{skin}). The three tests were done in a warm, dry environment (dry bulb temperature = 30°C; dew-point temperature = 11°C, equivalent to 30% relative humidity). Treadmill exercise ($1.36 \text{ m} \cdot \text{s}^{-1}$, 2% grade) was performed ($\sim 225 \text{ W} \cdot \text{m}^{-2}$; 425 W) for 80 min during each of the three experiments. The three cooling paradigms were presented in a balanced order. For each subject, experiments were done at the same time each day to avoid circadian effects on thermoregulation (21), but half the subjects were tested at approximately 10:00 and half were tested at approximately 12:30.

This experimental protocol, used previously without supplemental cooling in our laboratory for encapsulated female and male participants, increased core temperature $1.4 \pm 0.3^\circ\text{C}$ and heart rate 95 bpm at 75 min of exercise with skin temperature averaging 36.5°C in women (15); and increased core temperature $1.7 \pm 0.3^\circ\text{C}$ and heart rate 81 bpm at 75 min of exercise with skin temperature averaging $36.5\text{--}37^\circ\text{C}$ in men. (2,3).

Experimental procedures: After arriving at the laboratory, subjects drank 120 ml of water and body mass was determined. No further fluid intake was allowed dur-

ing the experiment. Participants wore spandex shorts, cotton socks, and athletic shoes. Over the next 30 min instrumentation with heat flow/skin thermistors, electrocardiogram (ECG) electrodes, and a heart rate monitor (Polar Watch, Polar CIC Inc., Port Washington, NY) was completed. Core temperature (T_c) was measured by temperature telemetry using an ingestible temperature sensor (HQ, Inc., Palmetto, FL) swallowed the night before an experiment. This ingestible sensor continuously transmitted the core temperature to a receiver that logged the data every min (FitSense, Southborough, MA). In some cases, the temperature sensor was ingested approximately 1.5 h prior to testing because the sensor that was ingested the day before was excreted prior to testing.

After the subjects were fitted with instruments, they donned the LCG and PPE. Fully clothed body mass was recorded before the start of exercise. Local skin temperatures and core temperature (T_c) were measured every minute and heart rate (HR) was obtained every 5 min by heart rate telemetry which was backed up by an electrocardiogram (Cardiovit AT-6, Schiller AG, Switzerland). Metabolic rate (M) was measured for a 3–4-min duration by open-circuit spirometry (TrueMax 2400, Parvo Medics, Inc., Sandy, UT) during the last 5 min of exercise after removal of protective mask and hood.

Total flow rate to the LCG was $1.2 \pm 0.05 \text{ L} \cdot \text{min}^{-1}$ when the cooling system was operated constantly. During active cooling in the two pulsed cooling paradigms, total flow rate to the LCG was also $1.2 \pm 0.05 \text{ L} \cdot \text{min}^{-1}$. The cooling system was run using an independent power supply which allowed the measurement of current and voltage usage during each of the three cooling paradigms. Heat extracted (or cooling) was also calculated from the product of flow, specific heat and specific density of the perfused fluid, and the gradient between the inlet and outlet temperature. In this case, heat (H , in W) was gained by the perfused fluid as it passed through the cooling garment. The electrical power (P) consumed was also measured (W , electrical).

Calculations: Mean skin temperature (\bar{T}_{sk}) was calculated from the formula:

$$\bar{T}_{sk} = 0.07 T_{head} + 0.10 T_{upper\ back} + 0.10 T_{lower\ back} + 0.10 T_{chest} + 0.10 T_{abdomen} + 0.14 T_{forearm} + 0.19 T_{thigh} + 0.20 T_{calf} \quad (5).$$

Mean body temperature (\bar{T}_b) was calculated from the formula used by Gonzalez (6) to describe exercising men with a similar \bar{T}_{sk} as was observed with CC:

$$\bar{T}_b = 0.2\bar{T}_{sk} + 0.8T_c$$

The core-to-skin temperature gradient was calculated as the difference between T_c and \bar{T}_{sk} . Whole body sweating rate (M_s) was determined from Δ semi-nude body mass corrected for respiratory water loss and $\text{CO}_2 - \text{O}_2$ exchange (16) and sweat left in clothing. Evaporative sweat loss (E_{sk}) was calculated from (Δ semi-nude mass – Δ clothing mass).

Data were analyzed through min 75 of exercise. For tabulation, the equation $M_{net} = M - W - (E_{res} + C_{res})$ was used (5,7) to calculate dry heat loss through the

skin (M_{net}), where M is metabolic heat production, W is external work, and E_{res} and C_{res} are respiratory heat loss from evaporation and convection, respectively (5). Skin insulation (I_{sk}) was calculated as $T_{\text{c}} - \bar{T}_{\text{sk}}/M_{\text{net}}$ (5). In three subjects (S1, S4, and S7) the telemetry pill data were unstable periodically during the experiments so the T_{c} data were fit to the quadratic equation using Sigma Plot (SPSS Science, Chicago, IL) that best fit the data. The instability of the data was likely due to the movement of the sensor in the gut, an observation that we have made previously (14) and tried to prevent by having the participants take the pill the night before testing. As noted earlier, this precaution was only partially successful. The predicted T_{c} data were then used as these data reflected the known T_{c} response to exercise over time for each subject.

A one-way analysis of variance with repeated measures was performed on the following parameters: T_{c} , \bar{T}_{sk} , and HR at 75 min, the change in T_{c} , the change in HR, M_{s} , M , M_{net} , W , E_{sk} , LCG inlet temperature, LCG outlet temperature, LCG flow, time that LCG perfusate was circulated through the tubing, the T_{c} to \bar{T}_{sk} gradient at min 75, I_{sk} , heat extracted, electrical power consumed, the T_{c} to \bar{T}_{sk} gradient/heat extracted, and heat extracted/time of cooling. The Holm-Sidak method for pairwise multiple comparisons post hoc tests was applied when significant main effects were found. Holm-Sidak is a less strict post hoc comparison than Tukey's honestly significant differences and was the recommended post hoc comparison when a data set failed the normality test (as was the case for inlet and outlet temperatures for the LCG). Statistical significance was set at $p < 0.05$.

RESULTS

The mean (\pm SD) metabolic rate was similar during the three cooling paradigms, averaging $222 \pm 22 \text{ W} \cdot \text{m}^{-2}$ in PC_{skin} , $226 \pm 25 \text{ W} \cdot \text{m}^{-2}$ in PC, and $227 \pm 26 \text{ W} \cdot \text{m}^{-2}$ in CC (NS). External work (W) averaged $11 \pm 0.4 \text{ W}$ for all experiments. Pre-exercise T_{c} was not different among the three cooling paradigms ($37.0 \pm 0.3^{\circ}\text{C}$ in PC_{skin} , $37.0 \pm 0.2^{\circ}\text{C}$ in PC, and $37.1 \pm 0.3^{\circ}\text{C}$ in CC). Likewise, T_{c} at 75 min was not different among the cooling paradigms ($37.6 \pm 0.3^{\circ}\text{C}$ in PC_{skin} , $37.6 \pm 0.2^{\circ}\text{C}$ in PC, and $37.6 \pm 0.2^{\circ}\text{C}$ in CC). Heart rate averaged $124 \pm 10 \text{ bpm}$ in PC_{skin} , $120 \pm 9 \text{ bpm}$ in PC, and $117 \pm 9 \text{ bpm}$ in CC. The mean \bar{T}_{sk} was $33.9 \pm 0.2^{\circ}\text{C}$ during PC_{skin} , $33.1 \pm 0.5^{\circ}\text{C}$ during PC, and $32.0 \pm 0.6^{\circ}\text{C}$ during CC at min 75 of exercise ($\text{PC}_{\text{skin}} > \text{PC}$, CC; $\text{PC} > \text{CC}$; $p < 0.05$). The mean (\pm SD) T_{c} , \bar{T}_{sk} , HR, and perfusate flow pumped through the LCG for the eight subjects are shown in Fig. 1. The top line in each panel shows HR. In Fig. 1 the data also show the temperature gradient between T_{c} and \bar{T}_{sk} as depicted by the lines shading the area of the graph between the two measurements. The bottom line shows the average flow rate of perfusate circulated each minute for the three paradigms. It illustrates the pulsatile flow for PC_{skin} and PC (bottom and middle panel, respectively) and the constant flow for CC (top panel).

There were no differences in the change in heart rate during exercise or the change in core temperature during exercise among the cooling paradigms (Table I).

The T_{c} to \bar{T}_{sk} gradient was greater for CC than PC and PC_{skin} (Table I, $p < 0.05$). In addition, the T_{c} to \bar{T}_{sk} gradient was significantly greater for PC than PC_{skin} ($p < 0.05$). Total electrical power used by the LCG was lowest for PC_{skin} , intermediate for PC, and highest for CC ($\text{PC}_{\text{skin}} < \text{PC}$, CC; $\text{PC} < \text{CC}$; $p < 0.05$, Table I). This represents a power savings of 46% in PC_{skin} and a 25% savings in PC compared with CC.

In the current study, the change in body mass due to evaporative sweat loss during the experiments averaged $0.8 \pm 0.3 \text{ kg}$ or $10 \text{ g} \cdot \text{min}^{-1}$ in PC_{skin} , $0.7 \pm 0.2 \text{ kg}$ or $8.8 \text{ g} \cdot \text{min}^{-1}$ in PC, and $0.6 \pm 0.2 \text{ kg}$ or $7.5 \text{ g} \cdot \text{min}^{-1}$ in CC (NS). There were no significant differences in evaporative heat loss or evaporative heat loss through the skin among the cooling paradigms.

Table II shows LCG data averaged across the entire experiment. For CC, cooling was supplied by circulating colder perfusate ($p < 0.05$) at a sustained flow rate ($p < 0.05$), but with a smaller gradient between inlet and outlet temperatures ($p < 0.05$) for a longer time ($p < 0.05$) than for PC by the experimental design. The same comparison can be made between CC and PC_{skin} . The temperature gradient between inlet and outlet perfusate and the time that perfusate circulated was greater in PC than PC_{skin} ($p \leq 0.05$; Table II).

Analysis of the physical data from the LCG indicated that during active cooling (Table I), the LCG extracted significantly less heat during PC_{skin} ($146 \pm 30 \text{ W}$) than PC ($196 \pm 17 \text{ W}$, $p \leq 0.05$) or CC ($253 \pm 26 \text{ W}$, $p \leq 0.05$), and $\text{CC} > \text{PC}$ ($p \leq 0.05$). Although heat was transferred to the LCG when no "active" cooling occurred in both pulsed cooling paradigms, this value was not measured because there were no temperature sensors embedded directly in the cooling tubes of the LCG.

The average body insulation was significantly smaller for PC_{skin} ($0.020 \pm 0.003^{\circ}\text{C} \cdot \text{W}^{-1} \cdot \text{m}^{-2}$) and PC ($0.024 \pm 0.004^{\circ}\text{C} \cdot \text{W}^{-1} \cdot \text{m}^{-2}$) than for CC ($0.029 \pm 0.004^{\circ}\text{C} \cdot \text{W}^{-1} \cdot \text{m}^{-2}$, $p < 0.001$), with no difference between the two pulsed cooling paradigms. A different way to look at heat transfer from the body through the LCG interface is to calculate the change in the T_{c} to \bar{T}_{sk} gradient for each unit of cooling provided by the LCG for the cooling paradigms. This calculation shows that PC_{skin} better preserved the T_{c} to \bar{T}_{sk} gradient with cooling ($0.027 \pm 0.01^{\circ}\text{C} \cdot \text{W}^{-1}$) than did either PC ($0.024 \pm 0.01^{\circ}\text{C} \cdot \text{W}^{-1}$) or CC ($0.023 \pm 0.004^{\circ}\text{C} \cdot \text{W}^{-1}$) ($p \leq 0.05$).

DISCUSSION

The pulsed cooling paradigm based on mean skin temperature feedback is an improved alternative for LCG cooling in exercising men clothed in encapsulated PPE as gauged by multiple criteria: 1) cardiovascular strain, as measured by the change in core temperature and heart rate, was not different among the three cooling paradigms; 2) core temperature at the end of the experiment (min 75) was not different among cooling paradigms; and importantly, 3) electrical power required for cooling was significantly reduced in PC_{skin} compared with both PC and CC.

A primary factor that controls skin blood flow is skin temperature. In particular, the effects of local skin temperature in the zone of vasomotor regulation (8,20),

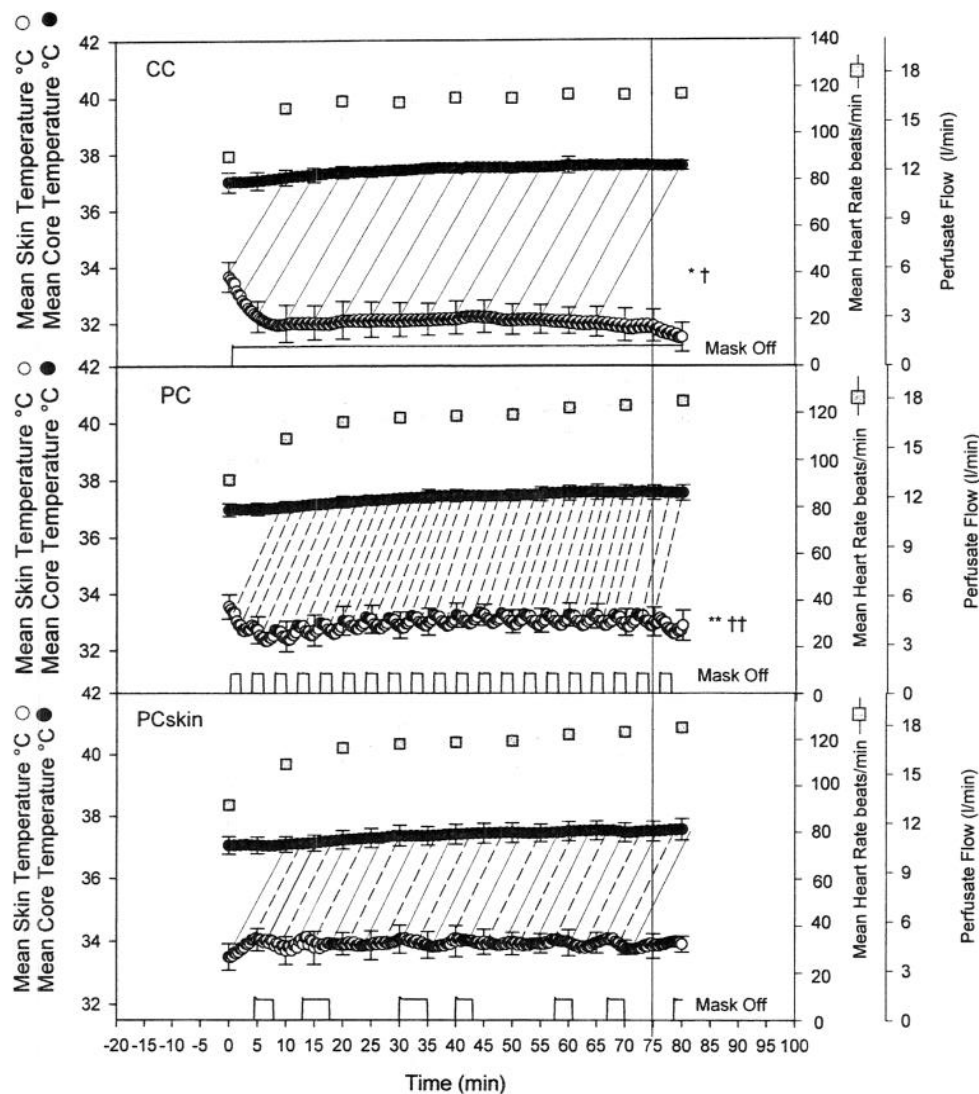


Fig. 1. Mean (\pm SD) heart rate, T_c , T_{sk} , and perfusate flow rate through the LCG for the three cooling paradigms. Constant cooling (CC) is shown in the top panel, pulsatile cooling (PC) is shown in the middle panel, and pulsatile cooling activated by skin temperature feedback (PC_{skin}) is shown in the bottom panel. The top line for each panel shows HR. The diagonal lines between T_c and T_{sk} for each cooling paradigm represent the T_c to T_{sk} gradient. The perfusate flow rate through the LCG is shown on the bottom line for each panel. For PC_{skin} perfusate flow rate is an idealized representation that best depicts the averaged T_{sk} response (bottom line). Note that the PPE encapsulation was compromised when the mask was removed after min 75 for the measurement of metabolic rate. T_{sk} : CC < PC, PC_{skin} $p \leq 0.05$; † perfusate flow rate: CC > PC, PC_{skin} $p \leq 0.05$; $^{**}\bar{T}_{sk}$: PC < PC_{skin} $p \leq 0.05$; $^{++}$ perfusate flow rate: PC > PC_{skin} $p \leq 0.05$.

TABLE I. PARAMETERS THAT CONTRIBUTE TO CARDIOVASCULAR STRAIN AND ENERGY USED IN COOLING.

	ΔHR (0–75 min) (Beats \cdot min $^{-1}$)	ΔT_c (6–75 min) ($^{\circ}C$)	T_c - T_{sk} gradient (at 75 min) ($^{\circ}C$)	Electrical Power (W)
Cooling Paradigm CC				
SD	26	0.49	5.7 $^{+}$	224 $^{+}$
	10	0.39	0.5	15
Cooling Paradigm PC				
SD	30	0.66	4.6 †	169 †
	9	0.23	0.6	16
Cooling Paradigm PC_{skin}				
SD	31	0.45	3.8	122
	10	0.27	0.4	18

Mean (\pm SD) change in heart rate (ΔHR), change in core temperature (ΔT_c) from min 6 to min 75, the T_c - T_{sk} gradient at 75 min and electrical power used.

*CC > PC ($p < 0.05$); $^{+}$ CC > PC_{skin} ($p < 0.05$); and † PC > PC_{skin} ($p < 0.05$)

TABLE II. MEAN INLET AND OUTLET TEMPERATURES TO THE LCG, FLOW RATE AVERAGED FOR THE FIRST 75 MIN OF THE EXPERIMENTS, THE DIFFERENCE BETWEEN INLET AND OUTLET TEMPERATURE AND THE TIME THE COOLING PUMP WAS ACTIVATED AND FLOW WAS "ON" TO THE LCG.

	Inlet Temperature (°C)	Outlet Temperature (°C)	LCG Flow Rate (L · min ⁻¹)	ΔT_{out-in} (°C)	Flow Activated (min)
Constant Cooling					
SD	21.5*†	24.4*†	1.2§**	2.9*	80§**
	0.4	0.5	0.0	0.5	-
Pulsed Cooling					
SD	22.4‡	26.0	0.6††	3.6††	40††
	0.7	0.8	0.0	0.5	-
Pulsed Cooling Controlled by Skin Temperature					
SD	24.2	26.9	0.5	2.7	32.5
	0.7	0.6	0.1	0.5	5.6

*CC < PC ($p < 0.05$); †CC < PC_{skin} ($p < 0.05$); ‡PC < PC_{skin} ($p < 0.05$); §CC > PC ($p < 0.05$);

**CC > PC_{skin} ($p < 0.05$), ††PC > PC_{skin} ($p < 0.05$)

activation of vasoconstrictor signal (8,10), release of vasoconstrictor activation (1,10,11,13,17,19,24), and frank vasodilation (10–12,23,24) have been well studied in humans and these studies were instrumental for PC_{skin} development. The key operational point with PC_{skin} was to maintain mean skin temperature between 0.5 and 1.5°C above the approximate thermoneutral temperature for the skin, i.e., 33°C (9). This mean skin temperature was chosen because it provided a reasonable gradient for heat flux through the skin, and it was above the skin temperature that would activate a vasoconstrictor signal in most resting humans. The design of the cooling paradigms effectively separated mean skin temperature into three significantly different values (Fig. 1).

The flexibility provided by physiological feedback, although reducing the T_c to \bar{T}_{sk} gradient during PC_{skin}, provides for the warmest \bar{T}_{sk} ($33.9 \pm 0.2^\circ\text{C}$) among the cooling paradigms and consequently a more vasodilated cutaneous vasculature during exercise. The greater cutaneous vasodilation during PC_{skin} resulted in greater radiative and convective heat flux, even though physical conduction was significantly less than during CC. The significantly reduced total body insulation led to better physiological efficiency as shown by improved T_c to \bar{T}_{sk} gradient per unit of heat extracted during PC_{skin} compared with PC and CC. By incorporating physiological efficiency in the PC_{skin} paradigm, significant electrical power savings were realized. Finally, the amount of heat extracted per min that cooling to the LCG was activated showed that PC_{skin} was significantly more efficient than CC.

The pulsed cooling paradigms improved heat transfer efficiency because heat was transferred to the liquid cooling garment as long as its temperature was below skin temperature even when the pump was not activated. The outlet temperature for PC was about 1.6°C higher than CC and this difference was even greater for PC_{skin} (2.5°C). Yet these perfusate temperatures were far below skin temperatures for each cooling paradigm, allowing for cooling during the non-active phases. Furthermore, the participants perceived PC_{skin} as acceptable as CC (22).

Activating cooling in a 2-min period, then deactivating cooling for the next 2-min period reduced body

insulation compared with CC, but this particular cooling activation cycle did not significantly improve the physiological efficiency as shown by improved T_c to \bar{T}_{sk} gradient per unit of heat extracted compared with CC. Nevertheless, PC significantly reduced electrical power required for cooling compared with CC by about 25%. It is interesting to note that PC was most efficient of the cooling paradigms when assessed by the heat extracted per min of cooling activation. This is particularly important because the cooling and deactivation periods can be readily altered for better physiological efficiency.

The physical data collected from the LCG during the cooling paradigms can easily be modified by several different approaches. In this study, LCG flow and the activation time for cooling were manipulated. In addition, the inlet temperatures can also be increased so that CC would have better physiological efficiency. That is, changing the perfusate temperature to one warmer than 21.5°C would also reduce vasoconstrictor activity in the cutaneous vasculature, which would decrease body insulation and increase radiative and convective heat flux. This approach would also result in electrical power savings and may be an acceptable alternative for light, but prolonged work. To optimize cooling for a given work intensity and environmental condition, LCG inlet perfusate temperature, LCG perfusate flow, and duration of cooling pulses can be altered.

Further modification of time-activated pulsed cooling can also be done to improve cooling efficiency. Xu et al. (25) have modeled various periods for cooling activation in PC. They showed that a better physiological efficiency can be attained during simulated work of a similar intensity as in the current study by cooling for a 5-min period, then deactivating cooling for the next 5 min with the cycle repeating. The 5-min period more closely represents the on/off cycles in the mean skin temperature feedback paradigm of the current study. Furthermore, Xu et al. have predicted the optimal PC paradigm for heavier work and for work done in hotter environments than the environment in the present study. Time-activated pulsed cooling makes it possible to set the degree of cardiovascular strain that is acceptable. It can be then used to predict work time given the time interval for cooling and deactivation of cooling in PC.

To summarize the current research findings, the critical element for optimizing cooling efficiency was to maintain mean skin temperature above the local skin temperature that activated cutaneous vasoconstriction. PC_{skin} significantly reduced body insulation, increased radiative and convective heat transfer, and improved the cooling efficiency. Clearly, the associated electrical power savings observed with PC_{skin} indicates that liquid cooling can be significantly adapted to optimize power consumption, yet still reduce cardiovascular strain as well as CC in individuals performing moderate work while clothed in PPE.

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